



Redesign of a Biomechanical Energy Regeneration-based Robotic Ankle Prosthesis using Indonesian Gait Data

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Highlights:

- Replication of calculation results based on SPARKy.
- Indonesia-specific robotic ankle design specification.
- Testing result of the Indonesian Robotic Ankle Control System.

Abstract. In this research, the robotic ankle design from Arizona State University (ASU) known as SPARKy was redesigned to accommodate the specific needs of Indonesian people. Most active prosthetic legs are designed based on gait parameters for people from Western countries, which may differ for people from other cultures that have a different anthropometry and economic background. Indonesians have smaller actuating power characteristics compared to people from Western cultures due to their smaller average weight and body height. Thus, the applied design strategy took advantage of a biomechanical energy regeneration scheme to reduce the actuator input power requirement and the relatively smaller mechanical power of the typical Indonesian ankle to create a potentially affordable robotic ankle with a smaller actuator that meets the technical specifications. The specifications of the powered prosthetic ankle were determined through the same methods used by SPARKy. Only one low-level control system, to actuate normal walking, was designed and tested on a fully assembled robotic ankle. The test results indicated a promising low-level control, where the robotic ankle can follow the predetermined trajectory required to actuate normal walking based on Indonesian gait data.

Keywords: *biomechanical; energy regeneration; prosthesis; robotic ankle; trajectory control.*

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1 Introduction

The desire to walk more naturally among people with a lower-limb amputation has motivated researchers to incorporate actuators to prosthetic designs, since intact ankles expend positive work to the environment in order to propel the user forward during walking [1,2]. Active prostheses have been widely developed across the world with various methods and control strategies [1,3]. To name a few, the Massachusetts Institute of Technology has developed a powered ankle prosthesis with a control scheme utilizing three low-level control layers, consisting of impedance, torque, and position control in every walking cycle [4]. The latest iteration by Carney, *et al.* [5] is a more lightweight, powerful and tunable robotic ankle, as opposed to a one-size-fits-all design. Cherelle, *et al.* [6] developed an active ankle prosthetic known as AMP Foot 2.0, which utilizes an elaborate mechanical system to emulate an intact ankle using a simple PI control loop. SPARKy from Arizona State University uses a simple PD motor position control loop implemented on their robotic tendon that uses an actuating method known as biomechanical energy regeneration to amplify the motor power density [7-9]. A lower-limb prosthetic with energy regenerative strategies has been developed by Khalaf, *et al.* [10], incorporating an impedance control system on the knee joint powered by a supercapacitor to make operation of the robotic leg last longer. The robotic ankle developed by Zhu, *et al.* [11], known as PANTOE, uses actuators on both ankle and toes to emulate walking. Bergelin & Voglewede [12] utilized the nonlinearities of a four-bar mechanism to emulate a sound ankle in their prosthesis design. It is worth noting that most powered ankle prostheses have one thing in common: they utilize an electromagnetic actuator coupled with planetary gearboxes, a rotational-to-linear transmission, and a spring. This configuration is called a series elastic actuator [13].

Although many studies have been done on different actuating methods and strategies for active lower limb prostheses, few have considered that different types of people may have varying anthropometries (which affects the power requirements) and economic background. The powered prostheses mentioned previously were mostly based on gait data from people living in Western areas such as North America or Europe [14-17]. Several researches have suggested that people from around the world may have different overall gait characteristics, which is worth considering when designing a prosthesis. For instance, Koreans have a considerably lower stride length and walking speed than subjects from Western countries [18]. Mahyuddin, *et al.* [19] have shown that the Indonesian gait also has unique anthropometric and spatio-temporal characteristics, where Indonesian people have a shorter stride length (1.15 m) and slower cadence (110 steps/minute) compared to the range provided by Whittle [15]. Since Indonesian's average body weight (60 kg) is lower than that of Westerners, which is about 70-80 kg [20], the total load on their ankle is smaller. Thus, a prosthesis

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designed based on Western gait data may be overdesigned and unnecessarily costly for Indonesians.

In 2010, Indonesia was home to more than three million disabled people in terms of lower-limb dysfunctionality, dominantly caused by diseases and accidents. This number is expected to rise due to the increasing number of diabetic cases and traffic accidents involving motorcycles, where most of the victims come from a poor economic background. Hence, more than 75% of Indonesians who may benefit from a prosthetic device are unable to access one [21,22]. There are commercially available prosthetic legs for amputees in Indonesia [23], however, these are passive, which causes mobility issues to amputees, such as increase of metabolic energy expenditure while walking, slower preferred walking speed, and pathological gait [2,4].

In this study, a robotic ankle prosthesis that suits the gait characteristics of a typical Indonesian was designed by redesigning an existing active prosthetic ankle. Preliminary work regarding the redesign has been previously presented in Sutawika [24] and Ferryanto, *et al.* [25]. The innovative low-cost passive Energy Storage and Return (ESAR) foot analyzed by Sugiharto, *et al.* [26] and Tazakka [27] was incorporated into the design to add a foot with better anthropometric resemblance to Indonesian people. The prosthetic ankle was manufactured and a low-level control system layer was developed to control normal walking mode. The control system and the device were assembled and tested in terms of the device's ability to follow the control command.

2 Methods

This research started by first understanding the robotic ankle SPARKy developed by Hitt [7] and discussed further in Hitt, *et al.* [8-9], including the working principles and methods used in its design. A calculation program was developed based on the equations and methods used for SPARKy. The calculation program was validated by comparing the resulting output specification values and plots generated by the program with the ones from Hitt, *et al.* [8-9]. If the results were similar, the design for an Indonesian robotic ankle could apply the same calculation program, using available Indonesian gait data instead. After obtaining the optimal specifications with the program, the components, control systems, and mechanical components were procured and designed based on these specifications. The robotic ankle was then manufactured, assembled, and tested. The testing of the robotic ankle in this research was limited to kinematic testing using a motion capture system.

2.1 Review of SPARKy Design

The reason why this study chose to redesign SPARKy for Indonesian amputees was because it has a simple mechanical design and control scheme, which makes it an attractive choice compared to other robotic ankles. A schematic of SPARKy is shown in Figure 1. The design utilizes a unique actuating method known as biomechanical energy regeneration. In this strategy, negative work is stored in the spring during the stance phase as the leg rotates about the ankle. A motor stretches the spring to add energy. The total energy stored in the spring is released during push-off, which produces similar kinetic and kinematic behavior as a sound ankle.

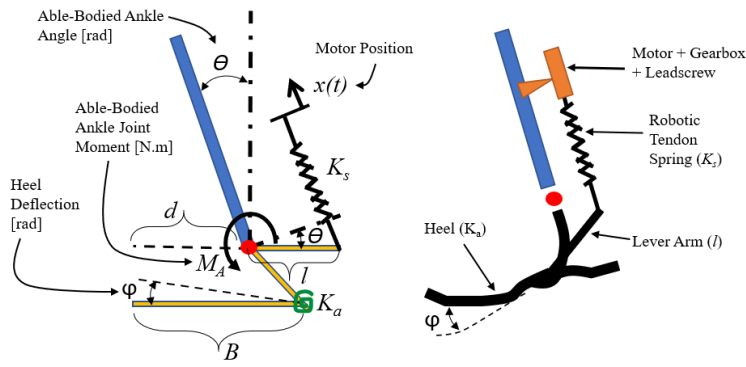


Figure 1 Schematic of SPARKy, redrawn from [7-9].

A robotic ankle utilizing biomechanical energy regeneration does not require a motor with a continuous power of 250 W (typical ankle peak power requirement during push-off for Western people), which may weigh around 6 to 7 kg. Using this approach, a smaller motor can be used. This is a great advantage in terms of power consumption and overall device weight. A very important achievement from the designers of SPARKy is that for a required peak power of 250 W to actuate a normal gait only a maximum of 80 W of instantaneous peak power is required from the motor to power the prosthesis. This is an apparent 3.25 power amplification from the motor input to the total prosthesis output. The other 170 W of power is obtained from the release of energy from the spring of the mechanism. The SPARKy design strategy seeks to minimize the required electrical power of the system, which can be calculated as follows:

$$P_e = \left[(J_m + J_g) \frac{d\omega}{dt} \right] \omega + \frac{\frac{M_A(t)}{l} [l\theta_A(t) - \frac{K_a B d}{K_s l} \phi(t)]}{\eta_m \eta_g \eta_{ls}} \quad (1)$$

where J_m and J_g are the inertia of the motor and the gearbox (kg.m^2); ω is the motor speed (rad/s); η_m , η_g , and η_{ls} are the DC motor, gearbox, and leadscrew

efficiencies, respectively; M_A is the ankle moment data [Nm]; θ_A is the ankle angle data [rad]; φ is the heel deflection [rad]; K_a is the heel stiffness [N/m]; K_s is the spring stiffness [N/m]; B is the radius of heel deflection [m]; d is the moment arm due to the heel deflection [m]; and l is the lever arm from the foot, which is attached to the nut of the leadscrew through the spring interface [m].

Eq. (1) is iteratively computed to obtain the robotic ankle parameters, consisting of the robotic tendon spring stiffness, gearbox ratio, leadscrew pitch, and lever arm length while minimizing the electric power input to the system. The determination of the final component parameters also considers the specifications for the DC motor. This means that the final selection of these parameter values influences the choice of DC motor. The complete explanation may be reviewed in either Hitt [7] or Hitt, *et al.* [8-9].

2.2 Indonesian Prosthesis Design

To redesign SPARKy, a calculation program was developed using the same equations to optimize the previously described robotic ankle parameters. The main inputs used to validate the calculation program were the kinematic and kinetic gait parameters (pentagram curve in Figure 2(a)). These gait parameters were digitized from the curve shown by Hollander, *et al.* [28], which was the input data used by Hitt, *et al.* [8] for their design. The aim of creating the calculation program was to first validate the equations used to create SPARKy. If the program output parameter values were similar to the ones produced in [7-9], the calculation program would be valid and could be used to iterate for the Indonesian robotic ankle. The gait parameters used as the input of the program is shown as the circle curve in Figure 2(a). Note that there are slight variations in the normalized gait parameters between Indonesians and Westerners. When the ankle moment of the average body mass of Indonesians and Westerners (60 kg and 80 kg) is compared, as depicted in Figure 2(b), Indonesians have a smaller peak ankle moment. This smaller ankle moment of the average Indonesian may be advantageous when designing an Indonesian robotic ankle, which could employ a smaller and less expensive motor.

Figure 3 shows the output curves that were generated by the developed calculation program based on the work of Hitt, *et al.* [8-9]. The slight differences between the plots in Figure 3 and in [8-9] may be due to the error from the digitization process of the gait parameters, assumptions of several values (such as the nut and leadscrew friction coefficient), and the apparently smaller number of data processed in the current work's program (each 2D plot had about 1000 data points). The results obtained confirmed that the calculation program developed for the current work was valid. Table 1 compares the values yielded

by the current work's calculation program to the values from [9]. Note that the two results are very close.

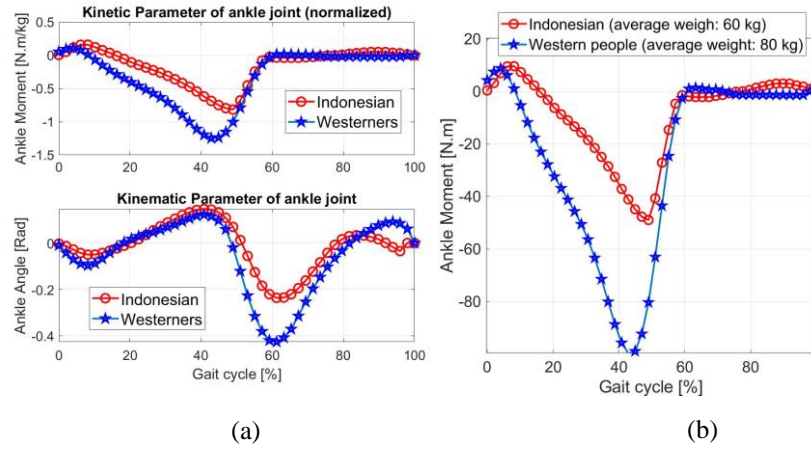


Figure 2 Comparison of gait parameters: (a) gait parameters for Indonesian and Western people [28], (b) comparison of ankle moment between Indonesian and Western people based on their respective average body masses.

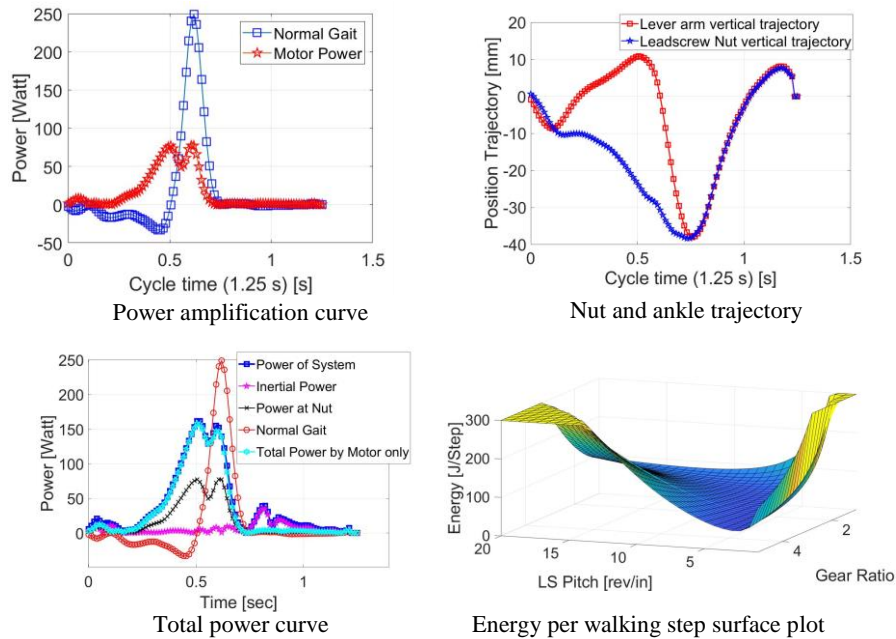


Figure 3 Results of the calculation based on the program developed in [9].

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Table 1 Comparison of output design values.

Calculated Parameters	<u>Optimal Parameter Calculation</u>	
	Hitt, <i>et al.</i> [9]	Current work
Gearbox ratio	1:4.6	1:5.0
Lead of leadscrew (rev/inch)	3	2.9
Lever arm length (m)	0.09	0.09
Spring stiffness (KN/m)	31.5	29.9
Power amplification (peak gait/peak actuator power)	3.25	3.20
Required electrical energy per step (Joules)	52.30	53.44

Table 2 summarizes the final iteration of the required robotic ankle parameters to develop an active ankle prosthesis that is compliant to Indonesian people with optimal values. Figure 4(a) shows that the average Indonesian weighs about 60 kg and only needs 65 W of maximum input mechanical power to the robotic ankle. This results in a virtual power amplification of 1.45. Even though the power amplification is not as high as that of SPARKy, the design reduces the power requirement of the ankle, which allows an even smaller actuator to be used. A 24-volt Maxon RE-32 brushed DC motor was selected for the current design. Figure 4(b) shows that this motor meets the design requirements because the total load on the motor due to normal walking is below the motor's speed-torque curve. The average power and torque per cycle exerted by the motor for a walking cycle time of 1.25 seconds are about 13 W and 0.0436 Nm, respectively. This is below the motor power rating and continuous maximum torque of 60 W and 0.0856 Nm, respectively. The calculated electrical energy from Eq. (1) expended per cycle is around 30 joules.

Table 2 Component specification for Indonesian robotic ankle prosthesis.

System Parameters	Result
Gearbox ratio	1:3.7
Lead of lead screw	8 mm/rev
Lever arm length	0.12 m
Spring stiffness	27.7 KN/m
DC motor	Maxon RE-32 24V

The mechanical design is based on SPARKy, with a few differences. The ESAR foot used in this work is based on previous research by Sugiharto, *et al.* [26], Tazakka [27], and Bawonoputro [29] and aims to create a low-cost prosthetic foot that is made entirely of aluminum but still has the same properties as the commercial carbon fiber foot characterized by Geil [30]. Figures 5(a) and (b) show a 3D model of the robotic ankle and the manufactured prototype, respectively.

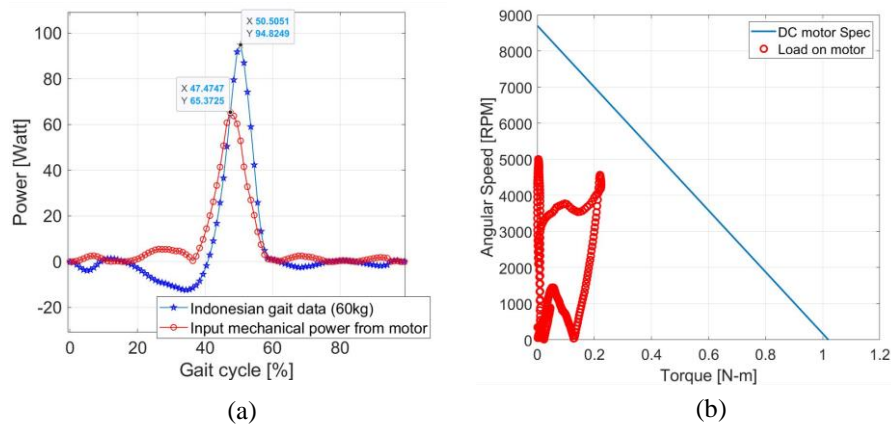


Figure 4 (a) Mechanical power at the ankle of the average Indonesian weighing 60 kg versus the input mechanical power required from the actuator to power the prosthesis, (b) load trajectory of normal walking compared to the selected motor specification.

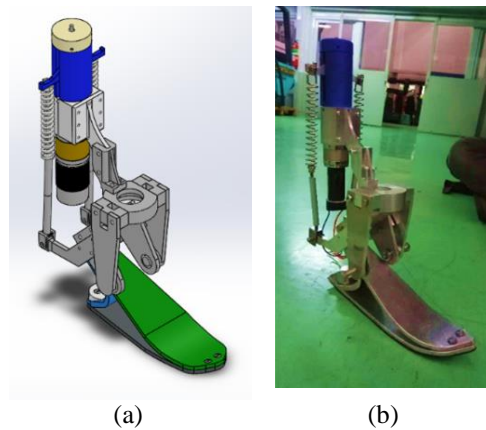


Figure 5 (a) Robotic ankle design and (b) manufactured device.

Ideally, a prosthesis/exoskeleton should have a control system with three layers, for high-, mid-, and low-level control, respectively [31]. In this research, only a low-level control layer was designed and implemented for the locomotive mode of normal walking. To implement biomechanical energy regeneration, the nut of the leadscrew (which is attached to the spring) must move in a predetermined vertical trajectory along the screw. As the nut moves, it stretches the spring in accordance with Hooke's law to generate the appropriate force on the lever of the prosthetic foot. Figure 6 shows the required vertical displacement of the nut to actuate normal walking for the current design based on the Indonesian gait as

shown in Figure 2(a), computed using the method in Hitt, *et al.* [8]. Figure 7 shows a diagram of the low-level control system for normal walking, where the nut's vertical position is controlled to follow the predetermined trajectory in Figure 6. The nut is actuated directly by the motor since the leadscrew is directly coupled to the gearbox of the DC motor. This control scheme follows the method from Ward, *et al.* [32]. In this study, the ability of the control system in Figure 7 to follow the trajectory given in Figure 6 was tested.

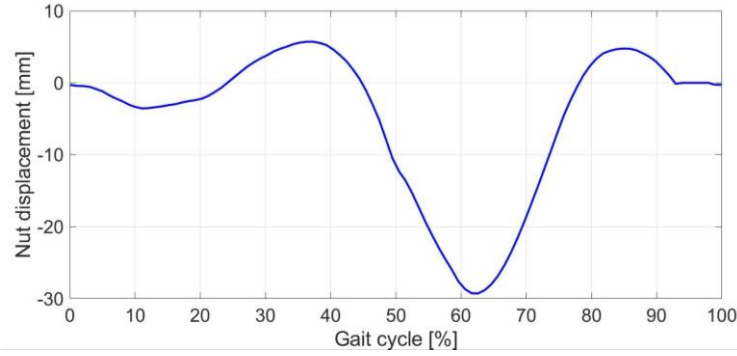


Figure 6 Required vertical nut trajectory.

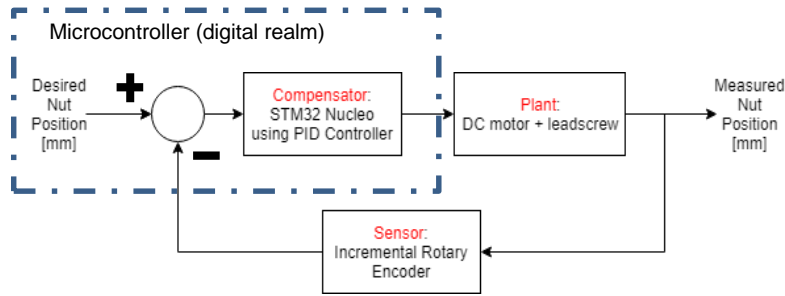


Figure 7 Low-level control system diagram for normal walking.

2.3 Testing of the Prosthesis

The testing of the robotic ankle in this research evaluated the developed low-level control layer for normal walking. The testing of the robotic ankle at this stage of the research only observed the kinematics of the nut and foot. In this case, the kinematics of the nut and the foot were the vertical displacement shown in Figure 6 and the ankle angle, respectively. Testing of the actual kinetic properties of the device was not conducted in this research. The experimental setup of the foot is shown in Figure 8(a), while Figure 8(b) shows the direction sign convention. The

series elastic actuator (SEA) shown in Figure 8(b) is a sub-assembly of the robotic ankle consisting of the motor and the transmission. A motion capture system was used to collect kinematic data of the nut and the ankle angle. The nut, ankle joint, toe, and heel were given LED markers to track their trajectories. When the foot is given the command by the microcontroller for normal walking, the motor moves the nut, and the nut moves the foot. The nut and the foot should have the exact same curve shape because the foot is suspended. The nut displacement and ankle angle data were collected directly through motion capture and indirectly through the DC motor's internal encoder. The captured data was later compared with the predetermined trajectory shown in Figure 6, which is also the command from the microcontroller.

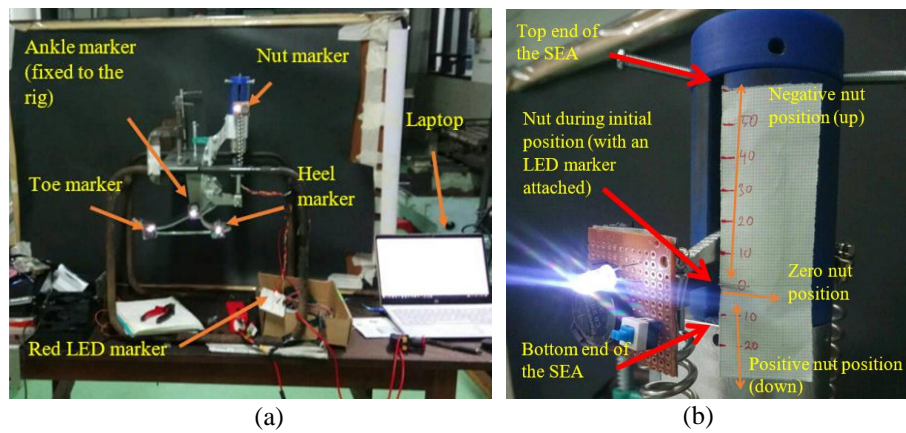


Figure 8 (a) Testing configuration and (b) direction sign convention.

3 Results and Discussion

Figures 9 and 10 show the captured data for the nut displacement and ankle angle, respectively. Figure 9 plots the setpoint/command from the microcontroller, the nut position read by the encoder, the nut position read by motion capture, and the error between the motion capture data and the command. The nut displacement curve read from the encoder and the motion capture system always overlap with the command. This means that the low-level control system was successful in creating the predetermined trajectory required to actuate a normal gait. The nut displacement read by the motion capture system mostly overlapped with the nut displacement command, which indicates a sufficiently good control result. The pentagram plot in Figure 9 shows the difference between the nut position from the motion capture and the command. The maximum difference was about 3 mm, which is acceptable. Even though there were discrepancies between the motion capture reading and the desired nut trajectory, the required vertical trajectory of

the nut was still successfully achieved and thus the control system design in this phase of the research was deemed satisfactory. Further research is required to test the control system under full operation in order to test the load capability of the electromechanical system.

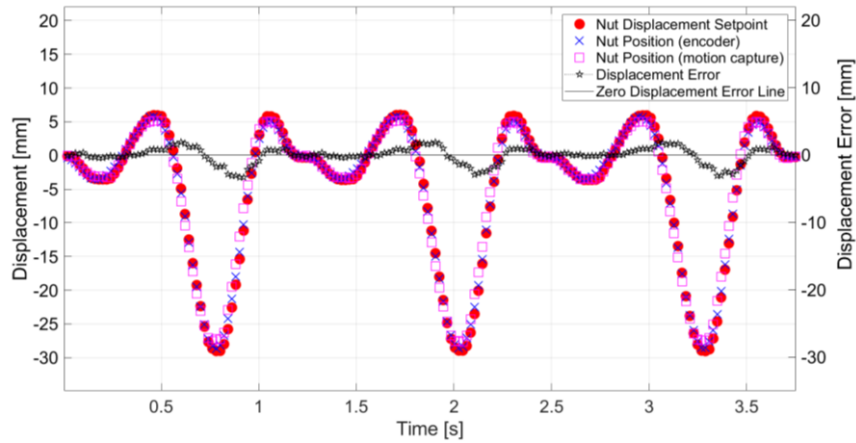


Figure 9 Nut displacement (setpoint, encoder, motion capture) and displacement error (motion capture to setpoint) versus time.

The ankle angle shown has a similar curve shape as the nut. Figure 10 shows the ankle angle of the robotic ankle plotted against the desired ankle angle of the normal Indonesian gait. During the stance phase there was a significant difference between the robotic ankle angle and the target ankle angle. This is reasonable because the foot was suspended on a rig. However, it can be seen that the angle of the robotic ankle slightly differed during the swing phase.

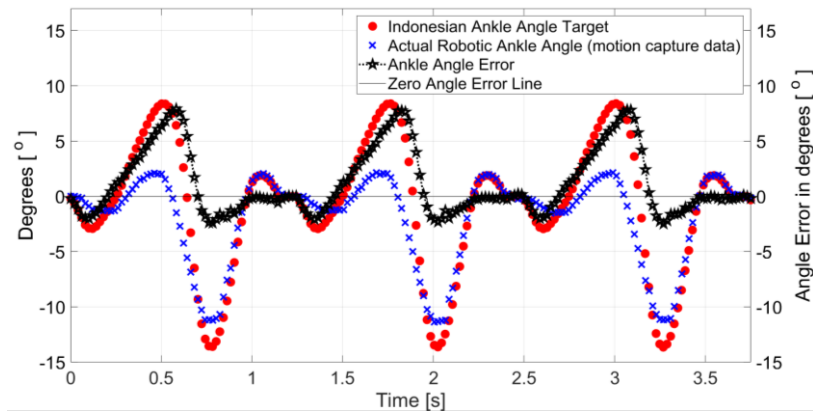


Figure 10 Robotic ankle angle, target ankle angle, and angle error (motion capture to target ankle angle) versus time.

It should be noted that, ideally, the robotic ankle angle should be the same as the required normal ankle angle during the swing phase (which spans from toe-off to heel-strike). This is because the foot is being positioned properly to prepare for the next gait cycle. However, the error between the robotic ankle and the required ankle angle was only about 2 degrees. This promising result shows that the prosthetic ankle has the potential to be further developed. Furthermore, it is clear from these results that the method used to develop SPARKy is reproducible from its design to its control strategy. Hence, this research has shown that the SPARKy method was successfully replicated and repurposed to fit another type of amputee.

The current device still lacks a controller to deal with detecting the state of the user, which is necessary before it is tested on a user. This controller also activates and modulates the low-level control layer to different walking speeds. Future work to improve the current design will include developing a control layer that may consist of a finite-state machine to determine when the user is about to initialize gait, increase walking speed, decrease walking speed, and stop gait as shown in Ward, *et al.* [32]. The mechanical design can also be improved to make a lighter device. Subject trials are necessary to evaluate the performance of the prosthesis. The design may also be compared to currently available prostheses.

Table 3 presents a comparison between the technical specification parameters of the current design and other powered prostheses. The first two technical specifications compared were the total mass and the sagittal-plane range of motion of the ankle prostheses. The other three specifications were the assigned power rating, maximum torque, and maximum speed of the actuator used to power each prosthesis. Comparing these parameters gives an idea of how redesigning a prosthesis based on a specific type of people may optimize the prosthesis further. Note that the BiOM Ankle Foot by Eilenberg, *et al.* [33,34] is the only powered ankle prosthesis that is available commercially.

Table 3 Comparison of technical specifications of powered prostheses.

Parameters	Powered Ankle Prosthetic Designs				
	Current work	SPARKy 1 [8]	Four-bar prosthetic [12]	AMP Foot 2.0 [6]	BiOM Ankle Foot [33,34]
Mass (kg)	2.3	Under 2.7	2.23	2.5	1.8
ROM ⁺ (Deg)	10-0-15	10-0-25	20-0-60	15-0-30	10.8-0-10.8
Actuator assigned power rating (W)	60	150	150	60	200
Max continuous actuator torque* (Nm)	0.0856	0.177	0.187	0.0517	0.120
Actuator speed at max continuous torque* (RPM)	8699	7575	7585	8169	16496

⁺) ROM: Range of motion, shown as dorsiflexion – neutral – plantarflexion.
^{*}) Condition of actuator without transmission.

It can be seen that most prostheses have a larger ROM and actuator power rating. This indicates that they are possibly overdesigned and in most cases are unaffordable for the average Indonesian. Even though they may have better specifications and could encompass a wider range of gait variations, the current work has shown that designing an active ankle prosthesis specifically for Indonesians is advantageous due to the smaller average kinetic and kinematic requirements. This enables the current design to use a smaller actuator compared to the original work. Hence, it is potentially more affordable compared to other comparable prostheses. It can be seen that the AMP Foot 2.0 is able to use a smaller actuator than the current work. However, the current work adopted SPARKy, which has a relatively simpler mechanical design compared to AMP Foot, which makes the current work preferable. In light of the current study, it may also be possible that if the AMP Foot design should adopt the same design principle proposed by this study, the AMP Foot may require an even smaller actuator.

4 Conclusion

A powered prosthetic ankle was designed, manufactured and tested. This research redesigned the robotic ankle known as SPARKy. A calculation program was developed using SPARKy's design principles and was used to design a prosthesis based on available Indonesian gait data. The robotic ankle's detailed component specification was determined using the calculation program. A low-level control layer for normal walking was developed and assembled to the device and a preliminary control system test was conducted by actuating the suspended foot on a rig. The results showed that the control system had satisfactory performance. The control system was able to position the nut into a predetermined trajectory that would actuate a normal gait. Further testing is required to evaluate the performance of the system in full load, including subject trials, to understand the full characteristics of the resulting design. The current design still has room for improvement such as optimizing the mechanical parts to be more lightweight and compact by using finite element analysis. Further research is needed to complete the control layers to enable adapting to changing conditions with respect to the user and the environment, and the ability to recognize and make transitions between different locomotion activities. A point worth noting is that the proposed design methodology can be employed to design robotic ankles based on gait data from different nationalities to potentially prevent overdesigning the device and cut overall cost, as shown in Table 3. Overall, this research has shown a successful replication of the design of SPARKy and validating the scheme of biomechanical energy generation via the developed iteration program.

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